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(54) Folded transverse gradient coil for magnetic resonance imaging systems

Gefaltete, transversale Gradientenspule für Anlagen zur Bildgebung mittels magnetischer Resonanz

Bobine à gradient transversal pliée pour systèmes d'imagerie par résonnance magnétique

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EP-A- 0 551 517 **WO-A-94/28430**
GB-A- 2 265 986

- **SOCIETY OF MAGNETIC RESONANCE IN MEDICINE (SMRM), ELEVENTH ANNUAL SCIENTIFIC MEETING, AUGUST 8-14, 1992, BERLIN, GERMANY, BOOK OF ABSTRACTS, VOL. 1, XP002021382 E.C.WONG ET AL.: "Short Cylindrical Transverse Gradient Coils Using Remote Current Return"**

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Description

[0001] This invention relates to magnetic resonance imaging systems.

[0002] In order to perform magnetic resonant imaging, it is necessary to provide a magnetic field gradient in spatial directions to encode locations of the sources of a magnetic resonance (MR) response signal. This encoded MR response signal is used to create the MR images. In conventional MR imaging systems there is a set of gradient coils through which current flows to produce a magnetic field which either adds or subtracts to a static magnetic field provided by a main magnet.

[0003] In order to create a magnetic field gradient, required for MR imaging, current must be provided over a surface with a specified current density. Since current requires a closed loop, there is a return loop which typically provides a magnetic field which is not the same as the desired magnetic field gradient. Typically the return loop distorts and weakens the desired gradient magnetic field.

[0004] The magnetic field gradient is intended to be created within an imaging volume inside the MR imaging apparatus. Stray magnetic fields which are produced outside of the imaging volume by the currents in the gradient coil can induce stray currents, known as eddy currents, in electrically conducting support structures in the magnet. The eddy currents produce additional and undesirable gradient fields within the imaging region. It is, therefore, desirable to contain the stray magnetic fields in order to restrict their ability to induce eddy currents.

[0005] One method of containing the stray magnetic fields is to provide a second layer of coils outside, and surrounding, the first layer of coils used to produce the magnetic field gradients. This second layer of coils produces a magnetic field of opposite polarity compared to that of the first layer, and neutralizes most of the magnetic field produced by the first layer of coils outside of the imaging volume. This requires driving current through an additional layer of coils. The power required to drive a current through coils is related to the strength of the desired gradient magnetic fields necessary to carry out a prescribed imaging process. Since large changes in current are desired to be applied over very short time periods, the power required can become very large. This, combined with the fact that there is a second layer of coils to drive, results in a large power requirement of the MR imaging apparatus.

[0006] Since the speed of the MR imaging, and the quality of the imaging data, is related to how quickly the current can be changed within the gradient coils, a two layer, or shielded, gradient coil design is inherently slower than the single layer design using the same coils.

[0007] The need for a return current makes the gradient coils longer, thereby causing some patients to become uncomfortable and claustrophobic when placed inside the imaging system.

[0008] Currently there is a need for an MR imaging system which contains stray magnetic gradient fields, also is faster for a given gradient amplifier power, and is not as confining as conventional magnetic resonance imaging systems.

[0009] It is an object of the present invention to meet these needs.

[0010] GB-A-2 265 986 discloses an X- or Y-gradient coil of an MRI apparatus. The coil comprises a folded spiral of an electrically conductive material, which includes a first set of loops disposed on an inner curved surface, e.g. a cylinder, and a second set of loops disposed on an outer curved surface. Selected loops of the outermost part of the first set of loops are connected to selected loops of the second set of loops by loop-connector parts extending between the curved surfaces at the fold.

[0011] Similar gradient coil devices are disclosed in EP-A-0 551 517, WO-A-94/28430 and in the Book of Abstracts of the eleventh SMRM meeting, 1992, page 583.

[0012] The invention provides a transverse gradient coil for use in an MRI system as defined in claim 1.

[0013] The invention provides a device for providing desired magnetic field gradients more quickly than conventional devices for the same gradient amplifier power.

[0014] The invention provides a device which requires lower gradient amplifier power to produce images at the same speed as conventional MR imaging devices.

[0015] The invention provides a magnetic resonance imaging device with a more open imaging region, thereby reducing the 'claustrophobic' effect on patients being imaged.

[0016] The invention provides a magnetic resonance imaging device which produces less acoustic noise as compared with conventional magnetic resonance imaging systems.

[0017] The invention provides a magnetic resonance imaging device which reduces the portion of the patient exposed to large magnetic field gradients to reduce the possibility of nerve stimulation.

[0018] Embodiments of the invention will now be described, by way of example, with reference to the accompanying drawings, in which:

[0019] Figure 1 is a simplified block diagram of a magnetic resonance imaging device in which the present invention may be utilized.

[0020] Figure 2 is an illustration of a patient positioned within a conventional transverse gradient body coil and main magnet of an MR imaging system.

[0021] Figure 3 is an illustration of a one of the 'fingerprint' coils of the transverse magnetic field gradient body coil of Fig. 2.

[0022] Figure 4 is an illustration of magnetic field lines of flux produced by the transverse magnetic field gradient body coil of Fig. 2.

[0023] Figure 5 is an illustration of a first embodiment of a transverse gradient coil design according to the present invention.

[0024] Figure 6 is a perspective illustration of one embodiment of the coil of present invention disposed upon two cylinders of radii a_1, and a_2.

[0025] Figure 7 is a elevational view of a gradient coil consistent with the present invention, used in connection with describing current flow.

[0026] Figure 8 is an illustration of a second embodiment of a transverse gradient coil design according to the present invention.

[0027] Fig. 1 is a simplified block diagram of the major components of a magnetic resonance (MR) imaging system suitable for use with the invention described herein. The system is made up of a general purpose mini-computer **2** which is functionally coupled to a disk storage unit **2a** and an interface unit **2b**. A radiofrequency (RF) transmitter **3**, signal averager **4**, and gradient power supplies **5a**, **5b** and **5c**, are all coupled to computer **2** through interface unit **2b**. Gradient power supplies **5a, 5b, 5c** energize gradient coils 12-1, 12-2, 12-3 to create magnetic field gradients G_x, G_y, G_z, respectively, in the "X", "Y", "Z" directions, respectively, over a subject 16 desired to be imaged. RF transmitter **3** is gated with pulse envelopes from computer **2** to generate RF pulses having the required modulation to excite an MR response signal from subject 16. The RF pulses are amplified in an RF power amplifier **6** to levels varying from 100 watts to several kilowatts, depending on the imaging method, and applied to an RF transmitter coil **14-1**. The higher power levels are necessary for large sample volumes, such as in whole body imaging, and where short duration pulses are required to excite larger MR frequency bandwidths.

[0028] MR response signals are sensed by a receiver coil **14-2**, amplified in a low noise preamplifier **9** and passed to receiver **10** for further amplification, detection, and filtering. The signal is then digitized for averaging by signal averager **4** and for processing by computer **2**. Preamplifier **9** and receiver **10** are protected from the RF pulses during transmission by active gating or by passive filtering.

[0029] Computer **2** provides gating and envelope modulation for the MR pulses, blanking for the preamplifier and RF power amplifier, and voltage waveforms for the gradient power supplies. The computer also performs data processing such as Fourier transformation, image reconstruction, data filtering, imaging display, and storage functions (all of which are conventional and outside the scope of the present invention).

[0030] RF transmitter coil **14-1** and receiver coil **14-2**, if desired, may comprise a single coil. Alternatively, two separate coils that are electrically orthogonal may be used. The latter configuration has the advantage of reduced RF pulse breakthrough into the receiver during pulse transmission. In both cases, the coils are orthogonal to the direction of a static magnetic field B_0 produced by a magnet means **11**. The coils may be isolated from the remainder of the system by enclosure in an RF shielded cage.

[0031] Magnetic field gradient coils **12.1**, **12-2**, and **12-3** are necessary to provide gradients G_x, G_y, and G_z, respectively, that are monotonic and linear over the sample volume. Multivalued gradient fields cause a degradation in the MR response signal data, known as aliasing, which leads to severe image artifacts. Nonlinear gradients cause geometric distortions of the image.

[0032] A conventional MR imaging transverse gradient magnetic field body coil, 12-1, 12-2 of Fig. 1, is shown in Fig. 2. It typically has four quadrants each which have a 'fingerprint' winding pattern 22, 24, 26, 28, similar to that shown in Figure 3. Current flows according to, or opposite arrows 21a, 23a, 25a, 27a. These quadrants are electrically connected in series with each other.

[0033] In each of the 'fingerprint' coils of Fig. 3, a surface current is designed to pass through a region 31 from A to B to cause a magnetic -field to be produced. This current path is designed to provide the desired magnetic field gradient. The region 33 from B to C is necessary to provide a current return path, completing the circuit. The return path in region 33, however, increases the energy stored in the magnetic field produced without providing a useful imaging gradient.

[0034] Figure 4 shows the lines of constant magnetic field component parallel to the main magnetic field for a steady current passing through the coils of the body coil of Figure 2, and winding pattern of Fig. 3. Points A, B and C of Fig. 4 correspond to A, B and C of Figure 3. As is evident in Figure 4, the gradient field produced by the return current path reduces the applied magnetic field gradient to zero at point C and actually reverses the gradient to the right of point C. Therefore, although the gradient is quite linear at the point A, it is reduced to an unusable level at point B and is, in fact, reversed beyond point C.

[0035] Since the magnetic field produced extends both into the cylinder and also outside of the cylinder, stray magnetic fields affect nearby equipment and objects. These magnetic fields are rapidly turned on and off, causing a time-changing magnetic field disturbance. It is therefore common to try to contain the stray magnetic fields.

[0036] A double layer gradient coil design is usually employed to contain stray magnetic fields outside of the gradient

coils. By adding a second cylinder of radius a_2 outside the inner cylinder of radius a_1 having a similar 'fingerprint' coils, but having the current in each 'fingerprint' coil running opposite that of a corresponding coil on the inner cylinder. The outer coils create a magnetic field gradient of opposite polarity as that produced by the inner gradient coil. This reduces the magnetic gradient field produced outside of the cylinders, but also reduces the field inside of the inner cylinder, which is the imaging volume, reducing the efficiency of the inner gradient coils.

[0037] Since the magnitude of a magnetic field created is inversely proportional to the radial distance from the current-carrying element, the effect of the inner cylinder on the imaging volume is significantly greater than that of the outer cylinder:

[0038] The magnetic field produced by a gradient coil is given by the Biot-Savart Law:

$$B = \frac{\mu_0}{4\pi} \oint_{coil} \frac{\vec{\lambda} \times \hat{r}}{R^2} dA \qquad (1)$$

where:

μ_0 is the magnetic field permeability in free space,
dA is an incremental current carrying element,
$\vec{\lambda}$ is the current density in the incremental current-carrying element dA;
$\hat{r}$ is a unit vector pointing in the direction from the incremental current carrying member to a point where the magnetic field is desired to be calculated; and
R is the scalar distance between the incremental current carrying member to the point where the magnetic field is being measured.

[0039] The current density $\vec{\lambda}$ determines the conductor path on each of the surfaces.

$$\vec{\lambda} = \lambda_\phi \hat{\phi} + \lambda_\rho \hat{\rho} + \lambda_z \hat{k} \qquad (2)$$

$$\lambda_\phi = c\sigma_\phi \cos\phi_0 \qquad (3)$$

$$\lambda_\rho = c\sigma_\rho \sin\phi_0 \qquad (4)$$

$$\lambda_z = c\sigma_z \sin\phi_0 \qquad (5)$$

[0040] The quantity c determines the overall strength of the surface current. The quantities σ_ϕ, σ_ρ and σ_z determine the shape of the current path and the properties of the resultant gradient field, such as its linearity. σ_ϕ is a measure of the number of windings per unit surface length. σ_ϕ may also vary across the length of the surface which the windings are disposed. In the simplest case σ_ϕ may be a constant. The quantities σ_ϕ, σ_ρ and σ_z can vary with the surface coordinates, ρ and z but not with ϕ. They are related to one another by the continuity equation:

$$\nabla \bullet \vec{\lambda} = \frac{1}{\rho_0} \frac{\partial \lambda_\bullet}{\partial \phi_0} + \frac{1}{\rho_0} \frac{\partial (\rho_0 \lambda_\rho)}{\partial \rho_0} + \frac{\partial \lambda_z}{\partial z_0} = 0 \qquad (6)$$

[0041] For an idealized transverse gradient coil, the magnetic fields produced by an inner cylinder of radius a_1 is:

$$B_1 = G_1^x x = \frac{\mu_0 c \sigma_\phi^{(1)}}{2a_1} \qquad (7)$$

where:

$\sigma_\phi^{(1)}$ is a function used to determine current density of the cylinder of radius a_1,
x is distance across the coil, and
G_1^x is the transverse magnetic field gradient produced by the inner cylinder in the x direction (across the cylinder).

[0042] Similarly, the magnetic field produced by an idealized transverse gradient coil on an outer cylinder of radius a_2 is:

$$B_2 = G_2^x x = \frac{\mu_0 c \sigma_\phi^{(2)}}{2a_2} \tag{8}$$

where:

$\sigma_\phi^{(2)}$ is a function used to determine current density on the cylinder of radius a_2,
G_2^x is the x transverse magnetic field gradient produced by the inner cylinder in the x direction.

[0043] Therefore the ratio of the magnetic fields gradients produced is:

$$\frac{B_2}{B_1} = \frac{G_2^x}{G_1^x} = \frac{\sigma_\phi^{(2)} a_1}{\sigma_\phi^{(1)} a_2} . \tag{9}$$

[0044] Therefore, the magnetic field strength is reduced by the effect of shielding. In addition, the return loop current, running in a different direction at the same radius, subtracts from the desired gradient magnetic field.

[0045] Since it is not necessary to include the return current path on the cylinder of radius a_1, and it has a smaller effect when at a larger distance, the return current path can be disposed on a surface a greater distance from the imaging volume, and have a smaller reducing effect on the magnetic field within the imaging volume. The return current may serve a second purpose, one of shielding to contain the stray magnetic fields without requiring a second full set of coils.

[0046] In Fig. 5, a single layer continuous gradient coil 50 is shown according to the present invention. It has a first region 51 having a plurality of half-loops 57 for carrying a current, a second region 53 having a plurality of half-loops 59 also for carrying a current, and a third region 55 having conductors which connect each half-loop 57 with a corresponding half-loop 59 to create a single coil 50. Coil 50 is intended to be folded or bent along the lines BB and B'B' to result in a shape which is disposed upon two cylinders of radii a_1, and a_2, as shown in Fig. 6. Section 51 is disposed upon a cylinder of radius a_1, while section 53 is disposed upon a cylinder of radius a_2. Section 55 is an intermediate used to link individual current paths of sections 51 and 53, connecting each turn of section 51 at radius a_1 to each corresponding turn of section 53 at radius a_2. This complication can be dealt with by appropriately soldering and supporting a connecting wire between each turn of coils at radius a_1 to those of radius a_2.

[0047] A pattern may be etched as indicated in Figure 5, on a single plane then folded approximately along the lines BB and B'B'. However, unless $(a_2-a_1)/a_1 << 1$, a pattern such as in Figure 5 cannot be folded and placed onto two cylindrical surfaces without buckling. It may be possible to design the region BB to B'B' using other than strictly linear patterns to permit sufficient puckering that the folding could be completed satisfactorily. Alternatively, it is possible to complete the region BB to B'B' with a discrete set of soldered jumper wires as mentioned above. In either case, these wires must be supported in a sufficiently strong fashion to withstand the forces caused by their interaction with the static magnetic field.

[0048] Section 53 serves two purposes - both to provide the necessary return current path, and to contain the fringing field. In order to accomplish these purposes most effectively, the two patterns will not be precise inverses of one another, but rather the outer path, section 53, may extend somewhat beyond the inner section 51 to effectively capture the flux at the ends.

[0049] Fig. 7 shows the geometry of the folded gradient coil. The surfaces 61, 65, 63 correspond to sections 51, 55, 53 of Figs. 5 and 6, respectively. The lengths z_1 and z_2, the angle θ which the coil is bent and the cylinder dimensions are marked. ϕ designates an angle about the central axis of the cylinders. ρ is a radial distance from the central axis. $\hat{\phi}, \hat{\rho}, \hat{k}$ are unit vectors in the directions of ϕ, ρ and the axis of the cylinders, respectively.

[0050] Since the dimensions and bending angle may be adjusted, relationships are set forth below for the current density for the coil to function properly.

[0051] Fig. 8 illustrates a second embodiment of the gradient coil pattern compatible with the present invention. In this embodiment, some return current paths may only be disposed upon section 55, and not reach to section 53. Section 53 may have fewer return loops than section 51, provided however, that the current densities are consistent with the following equations. These define the functions to determine surface currents needed to produce a desired magnetic fields according to Eq. (1):

[0052] Surface 61:

$$\sigma_\phi = \sigma_\phi^A \tag{10}$$

$$\sigma_z = \frac{z}{a_1}\sigma_\phi^A \tag{11}$$

where σ_ϕ^A defines the number of windings per unit length of surface 61. In other instances σ_ϕ, may vary according to z.

[0053] Surface 65:

$$\sigma_\phi = \sigma_\phi^B \tag{12}$$

$$\sigma_s = \frac{z}{\rho}\sigma_\phi^A + \frac{\rho - a_1}{\rho \sin\theta}\sigma_\phi^B \tag{13}$$

$$\sigma_\rho = \sigma_s \sin\theta \tag{14}$$

$$\sigma_z = \sigma_s \cos\theta \tag{15}$$

where σ_ϕ^B defines the number of windings per unit length of surface 65. In other instances σ_ϕ vary along surface 65.

[0054] Surface 63:

$$\sigma_\phi^C = -\left(\frac{z_1}{z_2}\sigma_\phi^A + \frac{a_2 - a_1}{z_2 \sin\theta}\sigma_\phi^B\right) \tag{16}$$

$$\sigma_z^C = \frac{z}{a_2}\sigma_\phi^C \tag{17}$$

where σ_ϕ^C defines the number of windings per unit length of surface 63. In other instances σ_ϕ vary according to z.

[0055] Therefore, by choosing the dimensions a_1, a_2, z_1, z_2, and angle θ, the current density $\lambda_\phi, \lambda_\rho, \lambda_z$ may be determined and conductor paths selected to produce these current densities required to produce the desired magnetic field. σ_ϕ^A, σ_ϕ^B and σ_ϕ^A may be adjusted to trade off between reducing stray magnetic field flux, increasing linearity of the gradient field, reducing the energy required to switch the gradient field and so on.

[0056] The precise winding density can also be slightly adjusted, as conventionally done in shim coil designs, to achieve a higher degree of magnetic field linearity, if desired.

[0057] The present invention has several major advantages over conventional transverse gradient coil designs.

[0058] First, the total stored energy required to achieve a given gradient is greatly reduced over conventional designs. This reduces substantially the power supply requirements that are a crucial consideration in the practicality of the design.

[0059] Second, the tendency of the return current to reduce the effective magnetic field strength will be eliminated thereby extending the region of useful gradient along the z axis (length of the cylinder).

[0060] Third, the length of the tube which the gradient coils are disposed could be shorter reducing the claustrophobic effect which patients experience.

[0061] Fourth, the shorter gradient coil tube reduces the portion of the patient subjected to large magnetic field gradients, reducing the possibility of nerve stimulation effects which may occur in large magnetic field gradients.
[0062] And fifth, a shorter gradient coil tube reduces the ringing and noise associated with MR imaging.

Claims

1. A transverse gradient coil (50) for use in an MRI system comprising:

a) a first surface (51, 61) comprised of a plurality of adjacent conductor half-loops (57) each capable of carrying a current, the first surface partially enclosing an imaging volume;
b) a second surface (53, 63) comprised of a plurality of adjacent conductor half-loops (59) each capable of carrying a current, the second surface arranged substantially parallel to the first surface, a distance away from the first surface, on a side of the first surface opposite that of the imaging volume;
c) a plurality of connecting conductors (55), connecting half-loops of the first surface to corresponding half-loops of the second surface such that a continuous coil is produced from the half-loops such that current of corresponding conductors of the first and second surfaces flows in opposite directions, characterized in that:

the conductors are formed to produce a surface current defined by:

$$\vec{\lambda} = \lambda_\phi \hat{\phi} + \lambda_\rho \hat{\rho} + \lambda_z \hat{k}$$

where

$$\lambda_\phi = c\sigma_\phi \cos\phi$$

$$\lambda_\rho = c\sigma_\rho \sin\phi$$

$$\lambda_z = c\sigma_z \sin\phi,$$

wherein the first surface (51, 61) has a partial cylindrical shape of radius a_1 and length z_1, and σ_ϕ, σ_z for the half-loops of the first surface are defined by:
$\sigma_\phi = \sigma_\phi^A$, where σ_ϕ^A defines the number of windings per unit length of surface and may vary along the first surface, and

$$\sigma_z = \frac{z}{a_1}\sigma_\phi^A;$$

wherein the second surface (53, 63) has a partial cylindrical shape of radius a_2 and length z_2, and σ_ϕ, σ_z for the half-loops of the second surface are defined by: $\sigma_\phi = \sigma_\phi^C$, where

$$\sigma_\phi^C = -\left(\frac{z_1}{z_2}\sigma_\phi^A + \frac{a_2 - a_1}{z_2 \sin\theta}\sigma_\phi^B\right),$$

and

$$\sigma_z = \frac{z}{a_2}\sigma_\phi^C;$$

and
wherein the surface (65) carrying the connecting conductors (55) makes an angle θ with the first surface, and σ_ϕ,

σ_ρ, σ_z for the connecting conductors are defined by: $\sigma_\phi = \sigma_\phi^B$, where σ_ϕ^B defines the number of windings per unit length of the connecting surface (65) and may vary along the connecting surface,

$$\sigma_s = \frac{z}{\rho}\sigma_\phi^A + \frac{\rho - a_1}{\rho \sin\theta}\sigma_\phi^B,$$

$$\sigma_\rho = \sigma_s \sin\theta,$$

and

$$\sigma_z = \sigma_s \cos\theta.$$

2. The coil of claim 1 wherein the half-loops of the second surface extend beyond those of the first surface so as to contain fringe magnetic fields from the half-loops of the first surface when activated.

3. The coil of claim 1 wherein the half-loops have an elliptical shape.

4. The coil of claim 1 wherein the half-loops and conductors are disposed on a single element which is folded to produce the first and second surfaces and the connecting conductors.

5. The coil of claim 1 wherein there are equal numbers of half-loops on the first and second surfaces, with each half-loop on the first surface being connected to a corresponding half-loop on the second surface.

Patentansprüche

1. Quergradientenspule (50) zur Verwendung in einem MRI System, enthaltend:

a) eine erste Oberfläche (51, 61), die von mehreren benachbarten Leiterhalbschleifen (57) gebildet ist, die jeweils einen Strom führen können, wobei die erste Oberfläche teilweise ein Bildgebungsvolumen umschließt,
b) eine zweite Oberfläche (53, 63), die von mehreren benachbarten Leiterhalbschleifen (59) gebildet ist, die jeweils einen Strom führen können, wobei die zweite Oberfläche im wesentlichen parallel zur ersten Oberfläche in einem Abstand entfernt von der ersten Oberfläche auf einer Seite der ersten Oberfläche gegenüber derjenigen des Bildgebungsvolumens angeordnet ist,
c) mehrere verbindende Leiter (55), die Halbschleifen der ersten Oberfläche mit entsprechenden Halbschleifen der zweiten Oberfläche verbinden, so daß eine durchgehende Schleife aus den Halbschleifen gebildet ist derart, daß Strom der entsprechenden Leiter der ersten und zweiten Oberflächen in entgegengesetzten Richtungen fließt,

dadurch gekennzeichnet, daß:
die Leiter so geformt sind, daß ein Oberflächenstrom erzeugt wird, der definiert ist durch:

$$\vec{\lambda} = \lambda_\phi \hat{\phi} + \lambda_\rho \hat{\rho} + \lambda_z \hat{k}$$

wobei

$$\lambda_\phi = c\sigma_\phi \cos\phi$$

$$\lambda_\rho = c\sigma_\rho \sin\phi$$

$$\lambda_z = c\sigma_z \sin\phi,$$

wobei die erste Oberfläche (51, 61) eine teilzylindrische Form mit dem Radius a_1 und der Länge z_1 hat und σ_ϕ, σ_z für die Halbschleifen der ersten Oberfläche definiert sind durch: $\sigma_\phi = \sigma_\phi^A$, wobei σ_ϕ^A die Anzahl der Windungen pro Längeneinheit der Oberfläche definiert und entlang der Oberfläche variieren kann und

$$\sigma_z = \frac{z}{a_1}\sigma_\phi^A;$$

wobei die zweite Oberfläche (53, 63) eine teilzylindrische Form mit dem Radius a_2 und der Länge z_2 hat und σ_ϕ, σ_z für die Halbschleifen der zweiten Oberfläche definiert sind durch:

$$\sigma_\phi = \sigma_\phi^C$$

wobei

$$\sigma_\phi^C = -\left(\frac{z_1}{z_2}\sigma_\phi^A + \frac{a_2 - a_1}{z_2 \sin\theta}\sigma_\phi^B\right),$$

und

$$\sigma_z = \frac{z}{a_2}\sigma_\phi^C;$$

und
wobei die die verbindenden Leiter (55) tragende Oberfläche (65) einen Winkel Θ mit der ersten Oberfläche macht und σ_ϕ, σ_ρ, σ_z für die verbindenden Leiter definiert sind durch: $\sigma_\phi = \sigma_\phi^B$, wobei σ_ϕ^B die Anzahl der Windungen pro Längeneinheit der verbindenden Oberfläche (65) definiert und entlang der verbindenden Oberfläche variieren kann,

$$\sigma_s = \frac{z}{\rho}\sigma_\phi^A + \frac{\rho - a_1}{\rho \sin\theta}\sigma_\phi^B,$$

$$\sigma_\rho = \sigma_s \sin\theta,$$

und

$$\sigma_z = \sigma_s \cos\theta.$$

2. Spule nach Anspruch 1, wobei die Halbschleifen der zweiten Oberfläche sich über diejenigen der ersten Oberfläche hinaus erstrecken, um so magnetische Randfelder von den Halbschleifen der ersten Oberfläche einzuschließen, wenn sie aktiviert sind.

3. Spule nach Anspruch 1, wobei die Halbschleifen eine elliptische Form haben.

4. Spule nach Anspruch 1, wobei die Halbschleifen und Leiter auf einem einzigen Element angeordnet sind, das gefaltet ist, um die ersten und zweiten Oberflächen und die verbindenden Leiter zu erzeugen.

5. Spule nach Anspruch 1, wobei sich gleiche Zahlen von Halbschleifen auf den ersten und zweiten Oberflächen befinden, wobei jede Halbschleife auf der ersten Oberfläche mit einer entsprechenden halbschleife auf der zweiten Oberfläche verbunden ist.

Revendications

1. Bobine (50) à gradient transversal destinée à une utilisation dans un système IRM comprenant :

a) une première surface (51, 61) constituée d'une pluralité de demi-boucles conductrices adjacentes (57) chacune capable de transporter un courant, la première surface renfermant partiellement un volume d'imagerie;
b) une deuxième surface (53, 63) constituée d'une pluralité de demi-boucles conductrices adjacentes (59) chacune capable de transporter un courant, la deuxième surface étant sensiblement parallèle à la première surface, à une certaine distance de la première surface, d'un côté de la première surface opposé à celui du volume d'imagerie;
c) une pluralité de conducteurs (55) de connexion, reliant des demi-boucles de la première surface et des demi-boucles correspondantes de la deuxième surface de sorte qu'une bobine continue est réalisée à partir des demi-boucles si bien que le courant des conducteurs correspondants des première et deuxième surfaces circule en sens opposés, caractérisée en ce que :

les conducteurs sont réalisés de façon à produire un courant de surface défini par :

$$\bar{\lambda} = \lambda_\phi \hat{\phi} + \lambda_\rho \hat{\rho} + \lambda_z \hat{k}$$

où

$$\lambda_\phi = c\sigma_\phi \cos\phi$$

$$\lambda_\rho = c\sigma_\rho \sin\phi$$

$$\lambda_z = c\sigma_z \sin\phi,$$

dans laquelle la première surface (51, 61) présente une forme cylindrique partielle de rayon a_1 et de longueur z_1, et σ_ϕ, σ_z pour les demi-boucles de la première surface sont définis par :
$\sigma_\phi = \sigma_\phi^A$, où σ_ϕ^A définit le nombre d'enroulements par unité de longueur de surface et peut varier le long de la première surface, et

$$\sigma_z = \frac{z}{a_1}\sigma_\phi^A;$$

dans laquelle la deuxième surface (53, 63) présente une forme cylindrique partielle de rayon a_2 et de longueur z_2, et σ_ϕ, σ_z pour les demi-boucles de la deuxième surface sont définis par :

$$\sigma_\phi = \sigma_\phi^C,$$

où

$$\sigma_\phi^C = -\left(\frac{z_1}{z_2}\sigma_\phi^A + \frac{a_2 - a_1}{z_2 \sin\theta}\sigma_\phi^B\right),$$

et

$$\sigma_z = \frac{z}{a_2}\sigma_\phi^C;$$

et

dans laquelle la surface (65) soutenant les conducteurs (55) de connexion forme un angle θ avec la première surface, et σ_ϕ, σ_ρ, σ_z pour les conducteurs de connexion sont définis par :
$\sigma_\phi = \sigma_\phi^B$, où σ_ϕ^B définit le nombre d'enroulements par unité de longueur de la surface de connexion (65) et peut varier le long de la surface de connexion,

$$\sigma_s = \frac{z}{\rho}\sigma_\phi^A + \frac{\rho - a_1}{\rho \sin\theta}\sigma_\phi^B,$$

$$\sigma_\rho = \sigma_s \sin\theta,$$

et

$$\sigma_z = \sigma_s \cos\theta.$$

2. Bobine selon la revendication 1, dans laquelle les demi-boucles de la deuxième surface s'étendent au-delà de celles de la première surface afin de contenir les champs magnétiques marginaux provenant des demi-boucles de la première surface lorsqu'elles sont activées.

3. Bobine selon la revendication 1, dans laquelle les demi-boucles sont de forme elliptique.

4. Bobine selon la revendication 1, dans laquelle les demi-boucles et les conducteurs sont placés sur un seul élément qui est plié pour produire les première et deuxième surfaces et les conducteurs de connexion.

5. Bobine selon la revendication 1, dans laquelle il existe un nombre égal de demi-boucles sur les première et deuxième surfaces, chaque demi-boucle de la première surface étant connectée à une demi-boucle correspondante de la deuxième surface.

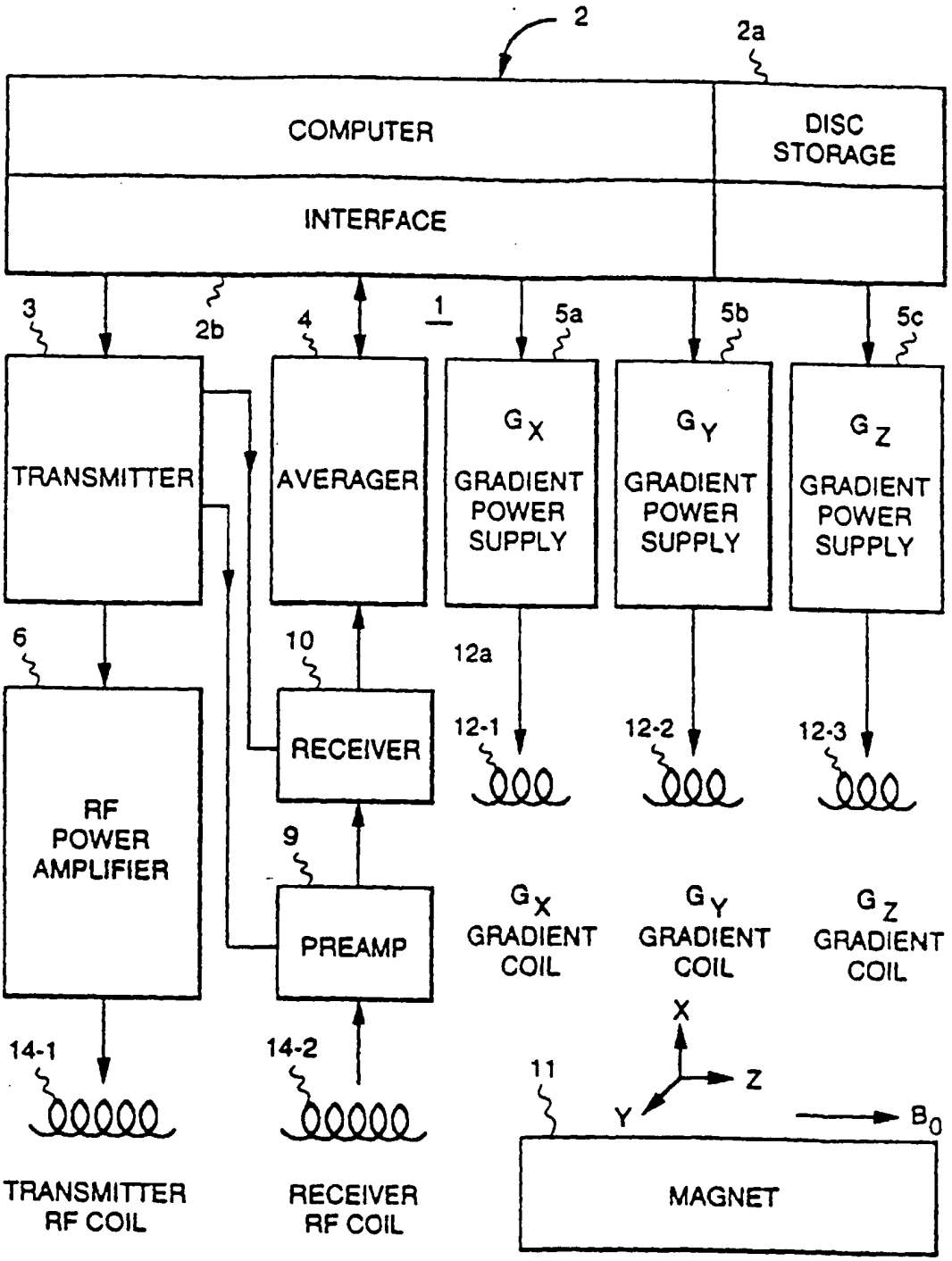
2
2a
COMPUTER
DISC STORAGE
INTERFACE
3
2b
4
1
5a
5b
5c
TRANSMITTER
AVERAGER
G_X GRADIENT POWER SUPPLY
G_Y GRADIENT POWER SUPPLY
G_Z GRADIENT POWER SUPPLY
6
10
12a
RECEIVER
12-1
12-2
12-3
RF POWER AMPLIFIER
9
PREAMP
G_X GRADIENT COIL
G_Y GRADIENT COIL
G_Z GRADIENT COIL
X
Z
Y
B_0
14-1
14-2
11
TRANSMITTER RF COIL
RECEIVER RF COIL
MAGNET

FIG. 1

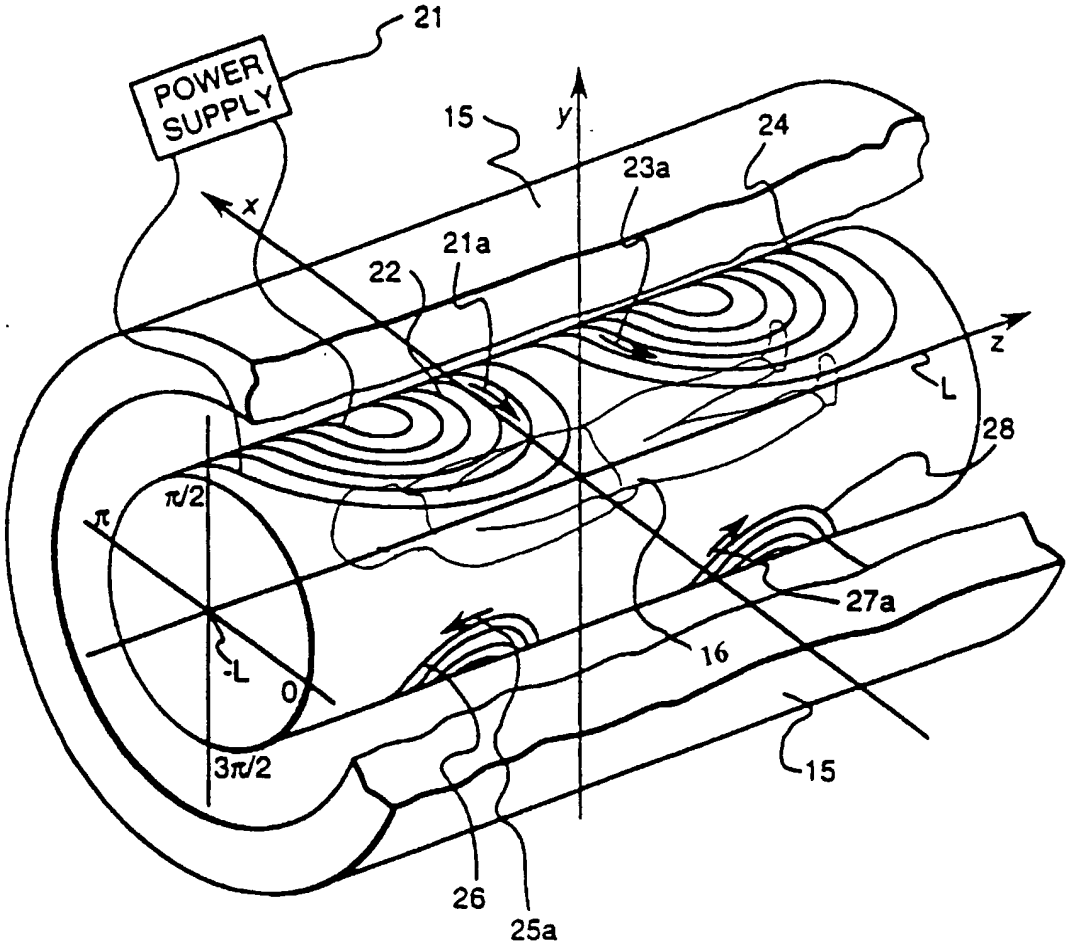

FIG. 2

PRIOR ART

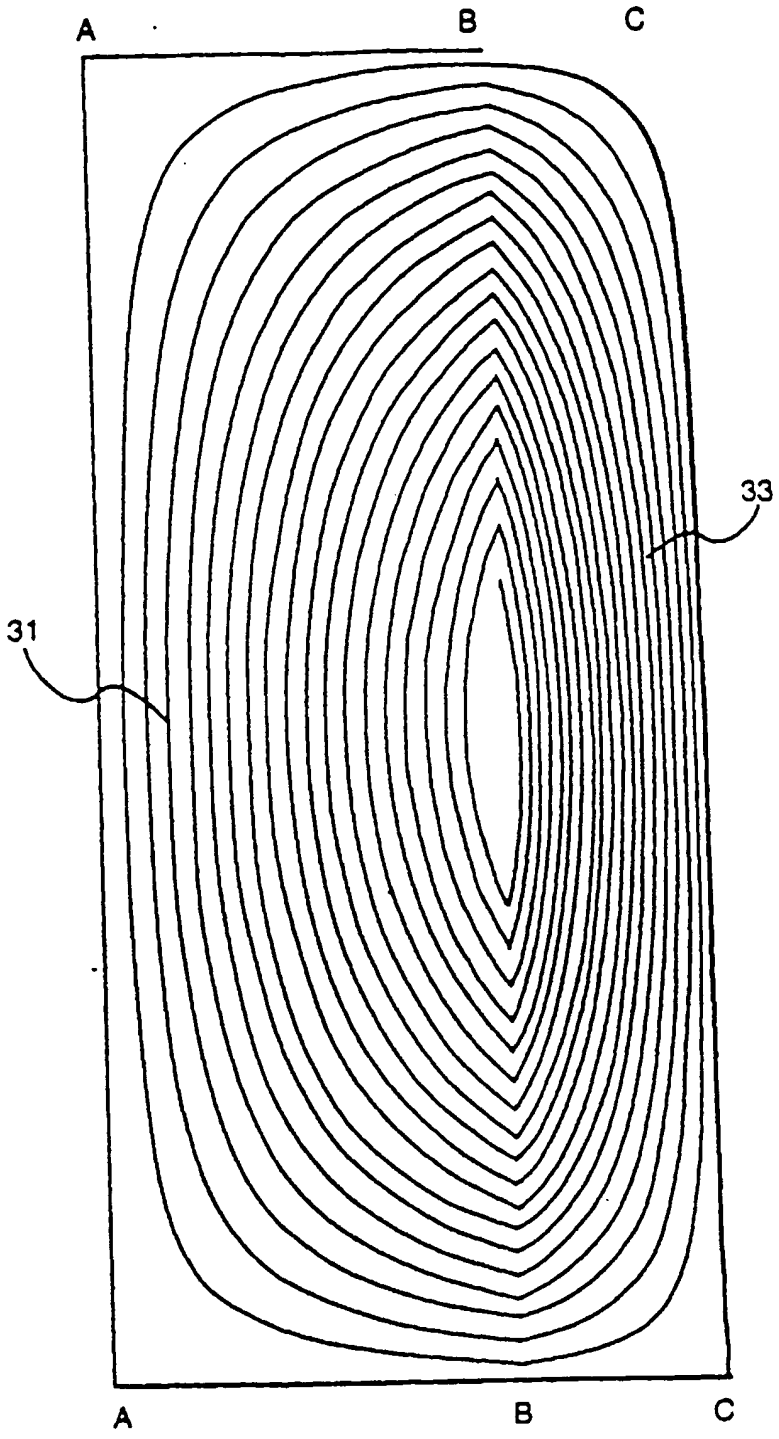

FIG. 3

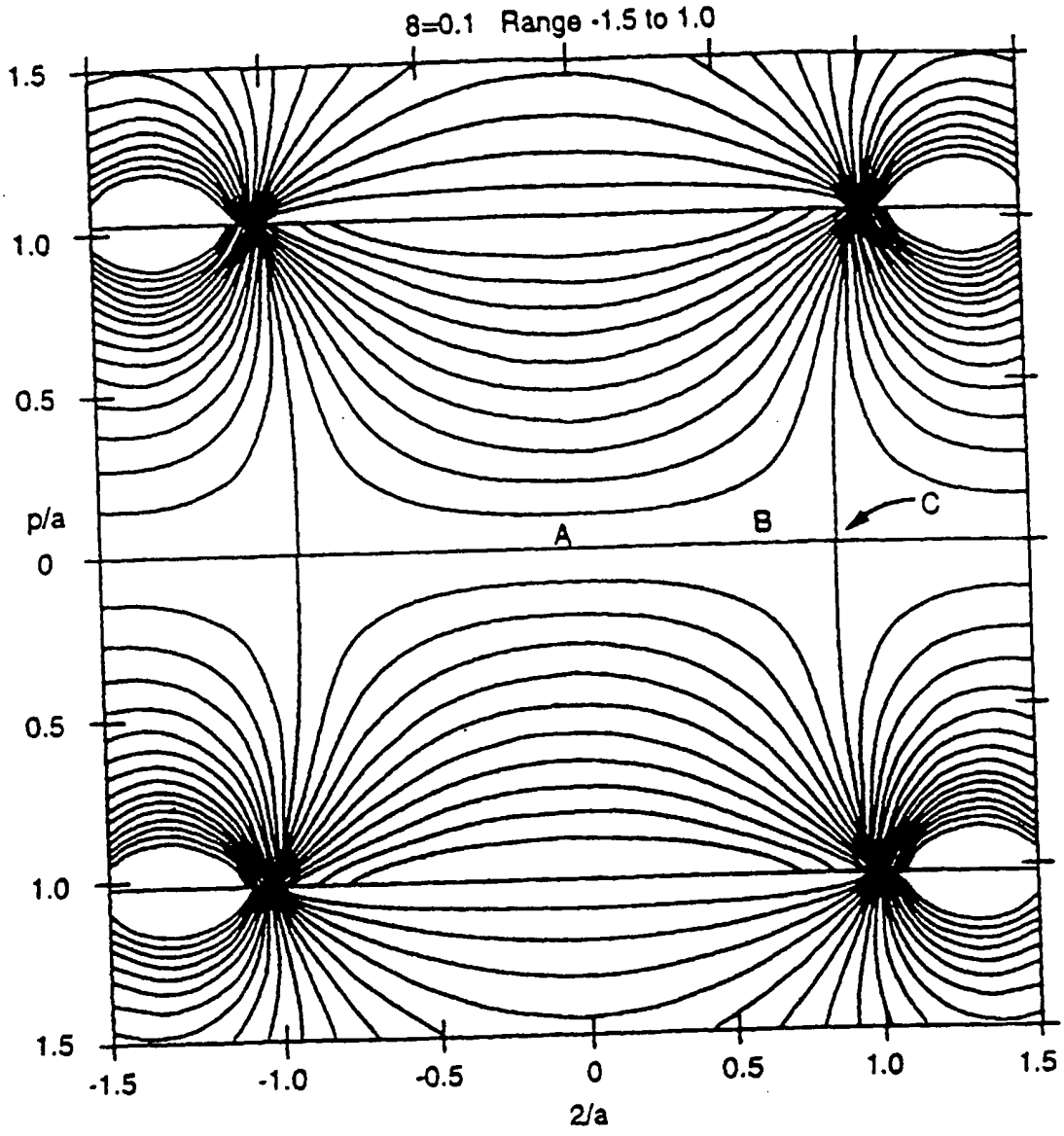

FIG. 4

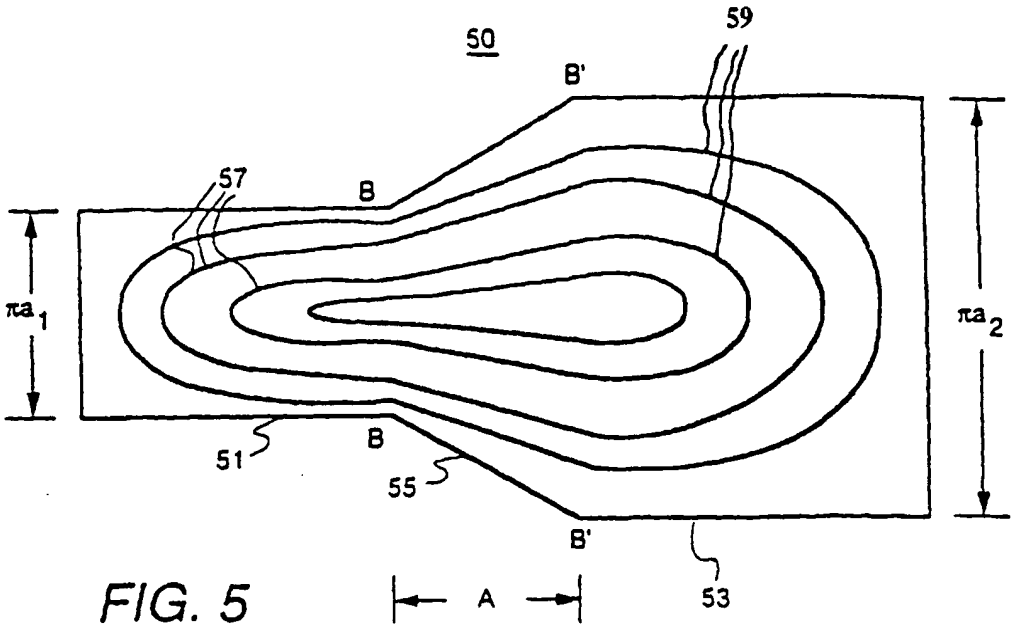

FIG. 5

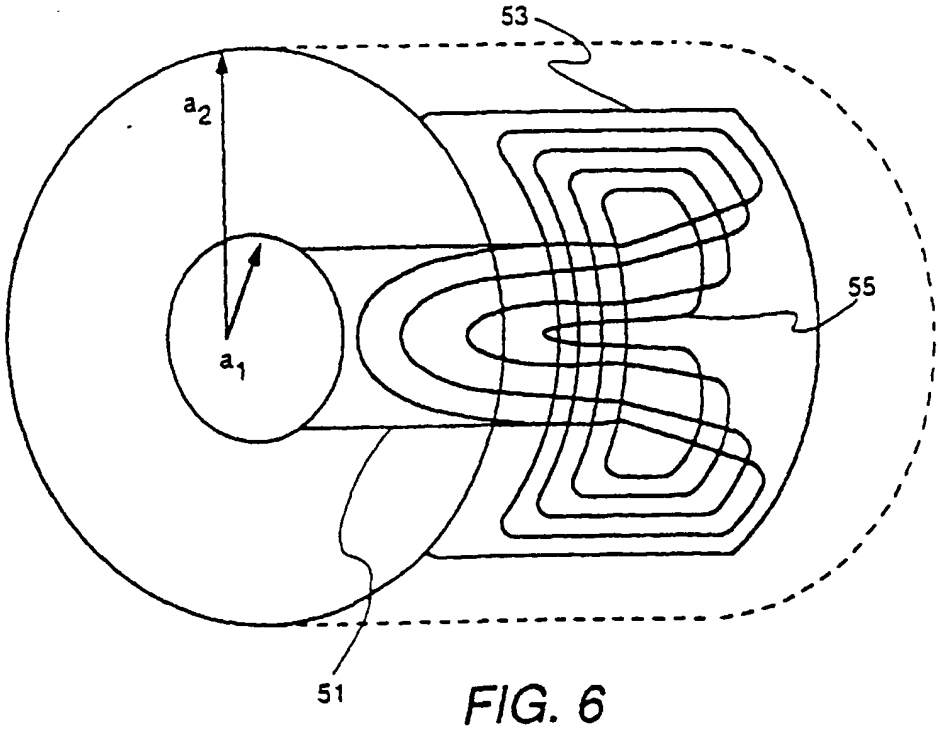

FIG. 6

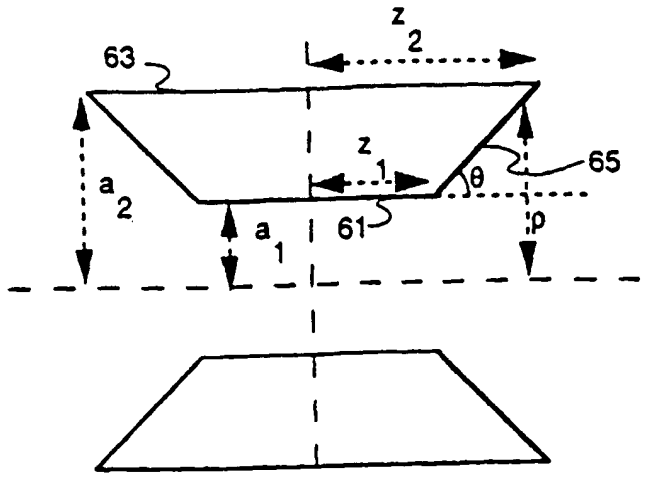

FIG. 7

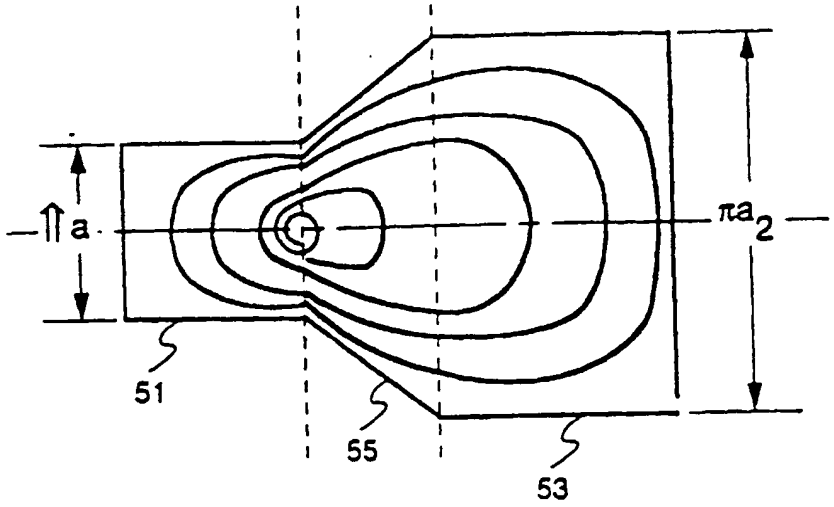

FIG. 8